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The purpose of this experiment was to investigate the effects of systematically manipulating the transverse plane moment of inertia (MOI) of the upper body on upper body torque, lower body torque, the phase relationship between the upper and lower body, and oxygen consumption. It was hypothesized that increasing the MOI of the upper body would result in (1) an increase in upper body torque that would be less than predicted solely from the increase in upper body MOI, (2) a decrease in lowerbody torque, (3) a more in-phase pattern of pelvic and thoracic rotation and, (4) an increase in oxygen consumption. Elevensubjects (4 male, 7 female, mean age: yr±SE = 26±2.0) walked on a treadmill at 1.3 ms-1 without a load and with an adjustablebackpack containing a load that was 40% of their body mass. Seven backpack MOI conditions were achieved by sliding two metalplates along a metal bar away from the transverse plane trunk axis of rotation. In the backpack condition with the smallestMOI, the predicted upper body torque was 1.6 times greater than the actual upper body torque, while in the largest backpack MOI condition, predicted upper body torque was 2.75 times greater than the actual. Increasing the MOI of the backpack resulted in no statistically significant change in lower body torque or oxygen consumption and a significantly more out-of-phase pattern of pelvic and thoracic rotation. We concluded that increasing the MOI of the backpack increases the reluctance of the upper body to changes in rotational movement. Consequently, changes in lower body torque and trunk coordination have less influence on upper body torque, and there is a releasing of movement degrees of freedom in the lower body.

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THE EFFECT OF BACKPACK MOMENT OF INERTIA ON TRANSVERSE PLANE KINETICS AND KINEMATICS AND OXYGEN CONSUMPTION DURING WALKING

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BACKGROUND

Previous load carriage research has shown that carrying a backpack (BP) results in increases in joint reaction forces (8, 17), oxygen consumption (11, 12, 13, 16), and changes in lower limb kinematics (6, 15). However, little research has been conducted to investigate the biomechanical mechanisms by which load carriage effects forces and torques, changes in movement patterns and metabolic cost. Recently, we have conducted experiments to investigate these mechanisms (9, 10). A major finding of this research was that transverse plane upper body torque was higher when carrying a BP than during unloaded walking; however, the upper body torque was about 45% less than would be predicted solely from the increase in upper body moment of inertia (MOI) caused by adding the BP (10). The reduction in upper body torque was achieved through a systematic decrease in transverse plane thoracic angular acceleration. The present study is focused on investigating the adaptations in gait that contribute to these changes in upper body torque and acceleration.

The findings of LaFiandra et al. (10) suggest that there are adaptations in the gait that prevent the generation of upper body torque during load carriage. It was found that transverse plane lower body torque was reduced by 50% during load carriage (10), which serves to decrease the lower body acceleration and, thereby, lower body torque. Reducing lower body torque reduces the amount of torque potentially transmitted to the upper body. It was also found that the net body torque (upper and lower body torque) during load carriage was greater than upper body torque, indicating that upper and lower body did not counterbalance each other as they do in unloaded walking.

The purpose of the present study was to further investigate the adaptations in transverse plane kinematics and kinetics due to changes in upper body MOI. Walking with a BP involves increases in mass and MOI. In order to isolate the effects of upper body MOI on transverse plane kinetic and kinematics, we systematically increased the upper body MOI without increasing the mass of the BP. This was achieved by sliding two metal plates along a metal bar away from the transverse plane trunk axis of rotation.

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LIST OF SYMBOLS

Symbol	Name
θ_{Pel}	Pelvic rotation (Deg)
θ_{Thor}	Thoracic rotation (Deg)
$\ddot{\theta}_{Pel}$	Pelvic angular acceleration (Deg $\cdot s^{-2}$)
$\ddot{\theta}_{Thor}$	Thoracic angular acceleration (Deg $\cdot s^{-2}$)
$T_{LBnetCCW}$	Net lower body counterclockwise torque (N· m)
$T_{LBnetCW}$	Net lower body clockwise torque (N· m)
$T_{LBPeakCCW}$	Peak lower body counterclockwise torque (N· m)
$T_{LBPeakCW}$	Peak lower body clockwise torque (N· m)
T_{Net}	Net torque (N· m)
$T_{UBnetCCW}$	Net upper body counterclockwise torque (N· m)
$T_{UBnetCW}$	Net upper body clockwise torque (N· m)
$T_{UBPeakCCW}$	Peak upper body counterclockwise torque (N· m)
$T_{UBPeakCW}$	Peak upper body clockwise torque (N· m)
ϕ_{cont}	Continuous relative phase (Deg)
ϕ_{disc}	Discrete relative phase (Deg)
VO_2	Oxygen consumption (mL/min)
VO_2 / BM	Oxygen consumption relative to subject's body mass (mL·kg·min)
VO_2 / TM	Oxygen consumption per total mass carried (body mass and BP mass) (mL·kg·min)
MOI	Moment of Inertia
IREDS	Infrared Light Emitting Diodes
VGRF	Vertical Ground Reaction Forces
ANOVA	Analysis of Variance
NBP	No Backpack
BP	Backpack

EXECUTIVE SUMMARY

The purpose of this experiment was to investigate the effects of systematically manipulating the transverse plane upper body MOI on upper and lower body torque, the phase relationship between the upper and lower body, and oxygen consumption. To determine the biomechanical mechanisms by which load carriage effects forces and torques, changes in movement patterns, and metabolic cost, 11 subjects walked on a treadmill at 1.3 $m \cdot s^{-1}$ without a load and with an adjustable BP containing a load that was 40% of their body mass. Seven BP MOI conditions were achieved by sliding two metal plates along a metal bar away from the transverse plane trunk axis of rotation.

It was hypothesized that increasing the upper body MOI would result in an increase in upper body torque that would be less than predicted solely from the increase in upper body MOI. In the BP condition with the smallest MOI, the predicted upper body torque was 1.6 times greater than the actual upper body torque, while in the largest BP MOI condition, predicted upper body torque was 2.75 times greater than the actual. Therefore, the results of this study support this initial hypothesis: the actual increase in upper body torque was less than predicted solely from the increase in BP MOI. As expected, the lower than predicted upper body torque was associated with systematic decreases in transverse plane thoracic angular acceleration.

Reducing lower body torque limits the amount of torque potentially transmitted from the lower body to the upper body. In this experiment, it was hypothesized that the lower body torque would decrease with increasing upper body MOI. In contrast, increasing the BP MOI resulted in no statistically significant change in lower body torque. Although the lower body torque was reduced compared to the no backpack (NBP) condition and was relatively small compared to upper body torque, it remained constant across all MOI conditions. We additionally hypothesized that increasing the upper body MOI would result in a more in-phase pattern of pelvic and thoracic rotation. The results did not support this hypothesis. A small but statistically significant increase in the phase relation was observed across increasing MOI conditions (from 68° to 97°). It was also hypothesized that, increasing the upper body MOI would result in an increase in oxygen consumption. In contrast, increasing the MOI of the BP resulted in no statistically significant change in oxygen consumption. The increase in upper body torque in the high BP MOI conditions apparently elicits a minimal increase in muscular force to control the load.

It was concluded that increasing BP MOI increases the reluctance of the upper body to changes in rotational movement. Consequently, changes in lower body torque and trunk coordination have less influence on upper body torque.

INTRODUCTION

Previous load carriage research has shown that carrying a BP results in increases in joint reaction forces (8, 17), oxygen consumption (11, 12, 13, 16), and changes in lower limb kinematics (6, 15). However, little research has been conducted to investigate the mechanisms by which load carriage effects forces and torques, changes in movement patterns, and metabolic cost. Recently, we have conducted experiments to investigate these mechanisms. A major finding of this research was that upper body (thorax, two arms; and head) torque in the transverse plane was higher when carrying a BP than during unloaded walking at all walking speeds (0.6 - 1.6 ms^{-1}). However, upper body torque was about 45% less than would be predicted solely from the increase in MOI caused by adding the BP (10). The less than predicted increase in upper body torque was achieved through a systematic decrease in angular acceleration of the upper body at all walking speeds. The present study is focused on the adaptations in gait that contribute to these changes in upper body torque and acceleration.

The findings of LaFiandra et al. (10) suggest that there are adaptations in the gait patterns that prevent the generation of upper body torque with the addition of a BP. It was found that transverse plane lower body (pelvis and two legs) torque was also reduced by 50% compared to unloaded walking (10). Reduction in pelvic rotation compared to unloaded walking served to decrease the lower body acceleration and, thereby, the torque. Limiting lower body torque reduced the amount of torque potentially transmitted to the upper body. It was also found that the net body torque (upper and lower body torque) was greater than upper body torque, indicating that lower body torque did not counterbalance upper body torque. This was achieved by maintaining a more in-phase pattern of transverse plane pelvic and thoracic rotation and of arm and leg swinging than normally seen during unloaded walking at higher walking speeds (19, 20, 21). LaFiandra et al. (9) found a more in-phase to a more out-of-phase pattern of pelvic and thoracic rotation emerged as a result of increasing walking speed while carrying a load. But, this phase change pattern observed with loaded walking was less than that observed during unloaded walking.

Wagenaar & Beek (20) suggest that at the higher walking speeds (> 0.8 ms^{-1}), the upper body torque counteracts the torque generated by the lower body. The question is whether the reverse can also be the case; that is, when carrying a load, can the lower body torque offset upper body torque and, therefore, decrease net body torque? The findings of previous research (9, 10) on the effects of carrying a BP during walking on kinematics and kinetics indicate that this is not the case. However, we did find that during loaded walking, ipsilateral arm and leg swings were more out-of-phase at the highest walking speed (20), suggesting that the high upper body torques at the higher walking speed may be attenuated by adaptations in gait that counterbalance the upper and lower body. These adaptations may emerge at higher walking speeds during load carriage than during unloaded walking. It may be the case that in previous load carriage studies, walking speed was not increased sufficiently to elicit the adaptations in gait that counterbalance upper and lower body torques while carrying a load.

The purpose of the present study was to further investigate the adaptations in gait preventing or counterbalancing the predicted increases in upper body torque due to the addition of load. Walking with a BP involves increases in mass and MOI. By systematically increasing the upper body MOI without increasing mass, a number of hypotheses can be generated. Since an increase in MOI potentially increases upper body torque, it may be proposed that the adaptations thought to reduce torque in the upper body would be brought into action to a greater extent when the MOI of the BP is large. Therefore, it was hypothesized that increasing upper body MOI would increase upper body torque, but not to the extent that would be predicted solely from the increase in MOI.

If the first hypothesis is supported, it would be possible to test the hypothesis that if there are adaptations in gait that prevent the generation of upper body torque, a systematic reduction in the lower body torque would be expected as upper body MOI increases. In contrast, if the role of lower body torque were to counteract the increased upper body torque, an increase in lower body torque would be predicted. If a goal of load carriage is to prevent the generation of upper body torque, the less-than-predicted increase in upper body torque is expected to be mediated by a systematic decrease in counter-rotation between the lower and upper body. The alternative hypothesis is that if the upper and lower body counteract each other, an increase in counter-rotation (a more out-of-phase pattern) between the pelvis and thorax would emerge as the upper body MOI increases. Furthermore, a counteractive strategy would predict that the net body torque is lower than the upper body torque.

If upper and lower body torques counteract each other, more muscular effort would be expected to be required to control the predicted increase in upper body torque which, in turn, may result in an increase in oxygen consumption. Therefore, it was hypothesized that walking with a BP that has a large transverse plane MOI will result in higher oxygen consumption than walking with a BP that has a small MOI.

METHODS

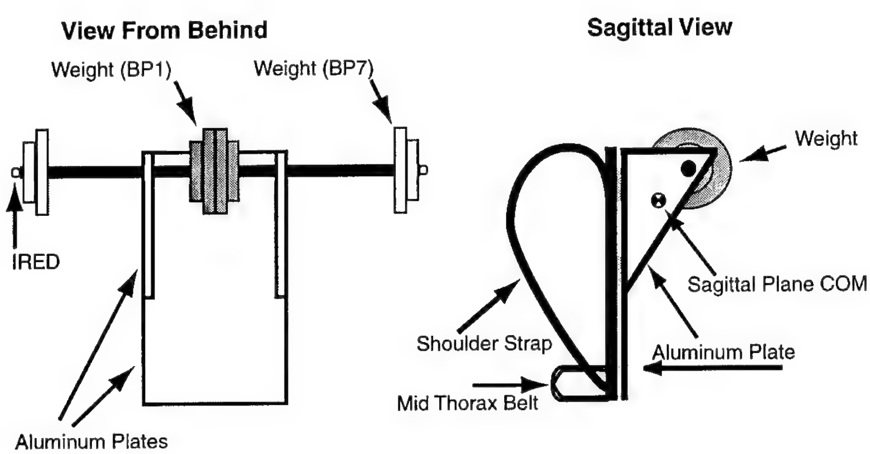
RESEARCH VOLUNTEERS

Twelve healthy subjects participated in the study. One subject was excluded from the analysis because of technical problems during data collection. The remaining seven female and four male volunteers were used in the final analysis. Subjects (age, yr: mean±SE = 26±2.0) were from the Boston University community, participated in strenuous physical exercise at least three times per week, and had no orthopedic disorders or complicating medical histories. Prior to participation, subjects gave informed consent in accordance with the policies of Boston University Institutional Review Board. The research was conducted in adherence with the provisions of 45 CFR Part 46.

EXPERIMENTAL BACKPACK

The BP frame was constructed of rigid plastic and designed to make contact only with the thorax (see Figure 1). An aluminum rod was attached to the frame to hold the weight at shoulder height as close to the subject's back as possible, and extended approximately 1 meter bilaterally, allowing for a manipulation of the BP MOI in the transverse plane. Two shoulder straps and a mid-thoracic strap minimized pack movement in relation to the thorax. The total weight of the BP was adjusted to 40% of the subject's body weight.

Figure 1. Illustration of backpack design to make contact solely with the thorax

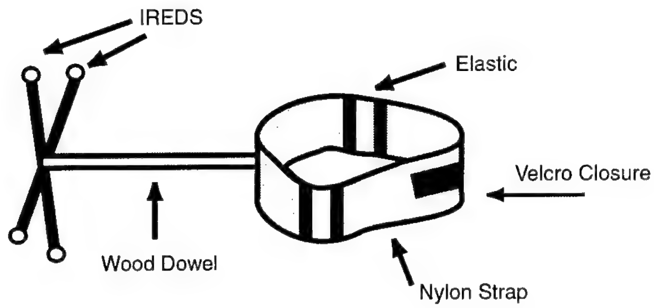


EXPERIMENTAL SETUP

Infrared Light Emitting Diodes (IREDS) were placed bilaterally on the subject's zygomatic processes, acromion processes, mid thighs, lateral femoral condyles, lateral malleoli, and ulnar styloid processes. Pelvic and thoracic motion were recorded using two custom made T-squares (20). IREDS were placed on each end of the T-squares

(Figure 2), bilaterally on the center of mass of the BP in the sagittal plane, and on the ends of the bar holding the weight.

Figure 2. Illustration of T square designed to track the movement of the pelvis and thorax



DATA COLLECTION

Data were collected in the Barreca Motion Analysis Laboratory at Boston University. Anthropometric measures were taken of total leg, shank, thigh, and arm length, as well as hip and shoulder width. Body mass and height were measured using a balance scale.

Three-dimensional kinematic data were collected at 100 Hz through an Optotrak 3020 System (Northern Digital Inc.). Cameras were placed on each side of a treadmill, approximately 3 meters from the treadmill. Subjects walked on an instrumented Kistler/Trotter treadmill instrumented with two force-plates capable of measuring vertical ground reaction force (VGRF). VGRF was sampled at 1000 Hz.

Oxygen consumption was measured using a breath-by-breath gas analysis system (Vacumed, Ventura, CA 93003, Vista Mini-CPX Silver Edition). The subject breathed into a face-mask that was connected to the gas analyzer. The gas analysis system incorporated an airflow meter, oxygen analyzer, carbon dioxide analyzer, and computer system. Oxygen uptake and carbon dioxide production were reported every 15 seconds.

Subjects walked on a level treadmill with NBP and with a BP at seven different MOI conditions (BP1 – BP7). In order to provide a baseline measure of oxygen consumption and transverse plane kinetics and kinematics during normal walking, the NBP condition was always the first condition collected. The sequence of BP MOI conditions (BP1 – BP7) was balanced across subjects. Subjects walked at 1.3 ms^{-1} for approximately 6 minutes in each condition. During the last 2 minutes, oxygen consumption data were collected. During the last 60 seconds, kinematic and kinetic data were collected.

DATA PROCESSING

Heel strike was determined to be the first frame of the time series in which the VGRF was greater than 7% of the peak VGRF (2). Each stride of data was time-normalized to percentage of stride.

The raw kinematic data were converted into three-dimensional data by means of the Optotrak system software. Missing data were interpolated using a cubic spline. If there were more than 15 consecutive frames of missing data within any stride, that stride was discarded because the interpolation was not reliable. The interpolation procedure was validated against known values before its use and demonstrated a maximum error of 1.0 mm. Aberrant force-plate data would occur if both feet were on the same force-plate at the same time. Strides with aberrant force-plate data were removed. There were an average of 46 strides (range 20-59) of data for each subject and condition. After interpolation, the data were filtered at 5 Hz (second order low pass Butterworth).

Transverse plane pelvic and thoracic rotations were measured from the filtered and interpolated time series. Continuous relative phase was calculated using the method described in van Emmerik and Wagenaar (9). Transverse plane pelvic and thoracic position and velocity data were normalized to minimal and maximal amplitudes (–1 and 1) within each stride. The phase angles for the pelvis and the thorax were calculated for every point in time. Continuous relative phase, ϕ_{cont}, between the pelvis and thorax is the difference between the pelvic and thoracic phase angles, and is calculated in the range of 0º-180°. Discrete relative phase, ϕ_{disc}, between pelvic and thoracic rotation was determined at heel strike.

Pelvic ($\ddot{\theta}_{Pel}$) and thoracic ($\ddot{\theta}_{Thor}$) angular accelerations were calculated as the second derivative of pelvic and thoracic rotation. The transverse plane axes of rotation of the pelvis and thorax were assumed to be located at the spine, midway between the greater trochanters of the left and right hips, and midway between the shoulders, respectively (18).

TORQUE CALCULATION

The MOI of the upper and lower body and torque variables were calculated using the method described in LaFiandra (10). MOI was calculated using Equation 1:

$$I = \sum_{i=1}^{n} m_i r_i^2 \qquad (1)$$

m represents the mass of the segment; r, the distance between the center of mass of that segment and the axis of rotation; and n, the number of segments. Segment mass and the position of each segment's center of mass were based on anthropometrics and calculated from estimates given by Dempster (3). The lower body included five segments: two shanks, two thighs, and the pelvis. The feet and shank were considered

one segment. The upper body included three segments: two arms and the thorax. The upper and lower arm were considered to be one segment. The mass of the head was included in the mass of the thorax segment. The contribution of the BP to the upper body MOI was calculated using the parallel axis theorem.

Net upper body torque was calculated separately in the clockwise and counterclockwise directions ($T_{UBnetCW}$ & $T_{UBnetCCW}$). Torque was calculated for each percentage of stride as the product of MOI and angular acceleration ($T = I\alpha$). $T_{UBnetCCW}$ is the sum of all the positive upper body torque, $T_{UBnetCW}$; is the sum of negative torque per stride. Upper body peak torque for each stride was also determined in both the clockwise and counterclockwise directions ($T_{UBPeakCW}$ & $T_{UBPeakCCW}$). Net and peak lower body torque ($T_{LBnetCW}$, $T_{LBnetCCW}$, $T_{LBPeakCW}$ & $T_{LBPeakCCW}$) were calculated in the same manner.

Predictions of upper body torque (T_{PRED}) were calculated for each BP MOI condition, as the product of upper body MOI (including the contribution of the BP) and thoracic angular acceleration in the NBP condition (Equation 2):

$$T_{PRED} = I_{UB}\ddot{\theta}_{ThorNBP} \qquad (2)$$

I_{UB} represents the upper body MOI, $\ddot{\theta}_{ThorNBP}$ thoracic angular acceleration in the NBP condition.

The net torque of the body (T_{Net}) represents the sum of all torque acting on the trunk per stride, calculated using Equation 3.

$$T_{net} = \sum_{i=1}^{100} \left| T_{LB_i} + T_{UB_i} \right| \qquad (3)$$

i represents the percentage of stride, T_{LB_i}, and T_{UB_i}, the torque of the upper and lower body for frame i.

OXYGEN CONSUMPTION CALCULATION

The average oxygen consumption was determined for the last 2 minutes of data collection. Oxygen consumption is expressed in absolute terms (VO_2 ml·min^{-1}), relative to the subject's body mass (VO_2 / BM ml·kg^{-1}·min^{-1}) and relative to the total mass carried (body mass and backpack mass, VO_2 / TM ml·kg^{-1}·min^{-1}).

STATISTICS

In order to determine differences between unloaded walking and walking with a load, an Analysis of Variance (ANOVA) with repeated measures was used to test for the

main effects of BP condition (two levels: NBP and BP) on the dependent variables. An ANOVA with repeated measures was used to test for the main effect of BP MOI (7 levels: BP1-BP7) on the dependent variables. If a significant main effect of BP MOI was found, Duncan Post-Hoc was used to determine the effect of increasing the BP MOI. An ANOVA with repeated measures was used to test for differences between T_{PRED} and the actual upper body torque. The significance level was set to $p < 0.05$ when testing from main effects. A Bonferroni correction was used to guard against Type I error in the post hoc analysis; significance was set to $p < .007$ (14)

RESULTS

Table 1 summarizes the P-values for the main effects of BP and the main effect of BP MOI for each of the dependent variables. Each variable is discussed in more detail in the following pages.

Table 1. P values for the main effects of BP, and for the main effect of BP MOI for each dependent variable.

	Main Effect of Backpack	Main Effect of Moment of Inertia
θ_{Pel} (deg)	0.0001	0.64
θ_{Thor} (deg)	0.0001	0.0001
$\ddot{\theta}_{Pel}$ (deg·s^{-2})	0.0529	0.4542
$\ddot{\theta}_{Thor}$ (deg·s^{-2})	0.0001	0.0001
$T_{LBnetCCW}$ (N·m)	0.0019	0.4150
$T_{LBnetCW}$ (N·m)	0.0018	0.4452
$T_{LBPeakCCW}$ (N·m)	0.0669	0.5802
$T_{LBPeakCW}$ (N·m)	0.0324	0.2439
T_{Net} (N·m)	0.0001	0.0001
$T_{UBnetCCW}$ (N·m)	0.0001	0.0001
$T_{UBnetCW}$ (N·m)	0.0001	0.0001
$T_{UBPeakCCW}$ (N·m)	0.0001	0.0001
$T_{UBPeakCW}$ (N·m)	0.0001	0.0001
ϕ_{cont} (deg)	0.0006	0.0001
ϕ_{disc} (deg)	0.0001	0.4597
VO_2 (mL/min)	0.0001	0.1467
VO_2 / BM (mL·kg·min)	0.0001	0.1592
VO_2 / TM (mL·kg·min)	0.0365	0.1583

Carrying a BP resulted in a statistically significant increase in counterclockwise net and peak upper body torque and clockwise net and peak upper body torque compared to unloaded walking (Table 2). A significant main effect of MOI was found for all upper body torque variables ($T_{UBnetCW}$, $T_{UBnetCCW}$, $T_{UBPeakCW}$, $T_{UBPeakCCW}$).

Table 2. Upper body torque: mean (standard error).

		Moment of inertia condition						
Variable	NBP	1	2	3	4	5	6	7
$T_{UBnetCCW}$ (N·m)	124.33* (1.34)	345.75^{F} (4.22)	338.99^{F} (4.00)	393.57^{E} (5.52)	545.84^{D} (10.27)	628.40^{C} (7.46)	727.82^{B} (10.16)	839.09^{A} (11.39)
$T_{UBnetCW}$ (N·m)	-125.31* (1.42)	-343.71^{F} (4.34)	-337.64^{F} (4.24)	-391.15^{E} (5.76)	-544.45^{D} (11.28)	-624.23^{C} (8.15)	-716.11^{B} (10.94)	-822.22^{A} (12.02)
$T_{UBPeakCCW}$ (N·m)	7.13* (0.09)	20.88^{F} (0.25)	21.18^{F} (0.27)	23.43^{E} (0.34)	30.37^{D} (0.56)	35.37^{C} (0.55)	41.77^{B} (0.81)	52.99^{A} (0.92)
$T_{UBPeakCW}$ (N·m)	-5.30* (0.07)	-14.62^{F} (0.20)	-14.19^{F} (0.20)	-16.48^{E} (0.29)	-22.71^{D} (0.69)	-25.55^{C} (0.43)	-30.80^{B} (0.56)	-36.61^{A} (0.66)

* Significant differences found between loaded and unloaded walking conditions.
Within MOI manipulation, conditions with same letter are not significantly different.

Carrying a BP resulted in a statistically significant decrease in transverse plane thoracic rotation and transverse plane thoracic acceleration compared to unloaded walking (Table 3). In addition, a significant main effect of MOI was found on thoracic rotation and acceleration; increasing MOI resulted in a decrease in thoracic rotation and acceleration.

Table 3. Transverse plane thoracic rotation and angular acceleration: mean (standard error).

		Moment of inertia condition						
Variable	NBP	1	2	3	4	5	6	7
θ_{Thor} (Deg)	7.51 * (0.08)	4.51^{A} (0.08)	4.03^{B} (0.08)	4.12^{B} (0.08)	4.02^{B} (0.09)	3.65^{C} (0.06)	3.12^{D} (0.06)	3.02^{D} (0.05)
$\ddot{\theta}_{Thor}$ ($Deg \cdot s^{-2}$)	463.41 (5.12)	280.95^{A} (3.54)	269.44^{B} (3.50)	230.33^{C} (3.29)	223.88^{C} (3.88)	193.31^{D} (2.92)	171.84^{E} (3.11)	168.13^{E} (2.77)

* Significant differences found between loaded and unloaded walking conditions.
Within MOI manipulation, conditions with same letter are not significantly different.

Carrying a BP resulted in a statistically significant decrease in counterclockwise net lower body torque and clockwise net and peak lower body torque (Table 4). However, no significant main effect of BP condition was found for counterclockwise peak lower body torque. No significant main effect of MOI was found on any of the lower body torque variables ($T_{LBnetCW}$, $T_{LBnetCCW}$, $T_{LBPeakCW}$, $T_{LBPeakCCW}$).

Table 4. Lower body torque: mean (standard error).

		Moment of inertia condition						
Variable	NBP	1	2	3	4	5	6	7
$T_{LBnetCCW}$ (N·m)	230.51* (3.15)	148.17 (3.05)	127.05 (2.45)	118.84 (2.20)	137.30 (3.77)	140.23 (4.01)	131.88 (2.98)	154.02 (4.76)
$T_{LBnetCW}$ (N·m)	-233.14* (3.16)	-147.66 (3.30)	-127.28 (2.44)	-118.12 (2.25)	-136.39 (3.74)	-139.34 (3.76)	-133.19 (3.19)	-156.03 (5.00)
$T_{LBPeakCCW}$ (N·m)	11.224 (0.19)	8.94 (0.21)	8.15 (0.20)	7.42 (0.16)	8.66 (0.26)	8.74 (0.27)	8.37 (0.23)	9.60 (0.31)
$T_{LBPeakCW}$ (N·m)	-11.43* (0.17)	-9.09 (0.24)	-7.32 (0.16)	-6.57 (0.15)	-8.14 (0.26)	-8.72 (0.29)	-7.96 (0.22)	-9.59 (0.36)

* Significant differences found between loaded and unloaded walking conditions.

Carrying a BP resulted in a statistically significant decrease in transverse plane pelvic rotation (Table 5). No significant main effect of BP condition was found for transverse plane pelvic acceleration ($p = 0.0529$). In addition, no significant main effect of MOI was found on either θ_{Pel} or $\ddot{\theta}_{Pel}$.

Table 5. Transverse plane pelvic rotation and angular acceleration: mean (standard error).

		Moment of inertia condition						
Variable	NBP	1	2	3	4	5	6	7
θ_{Pel} (Deg)	8.14 * (0.11)	4.76 (0.10)	4.26 (0.08)	4.107 (0.08)	4.25 (0.09)	4.30 (0.12)	3.92 (0.07)	4.46 (0.12)
$\ddot{\theta}_{Pel}$ ($Deg \cdot s^{-2}$)	452.98 (6.14)	343.50 (8.12)	314.12 (6.61)	280.87 (5.94)	316.35 (8.16)	320.00 (8.43)	312.35 (7.57)	360.91 (9.48)

* Significant differences found between loaded and unloaded walking conditions.

The net torque of the body (T_{Net}) represents the sum of all torque acting on the trunk per stride. Carrying a BP resulted in a statistically significant increase in net body torque (Table 6) compared to the NBP condition. A significant main effect of MOI condition was also found on net torque; increasing MOI resulted in an increase in net body torque.

Table 6. Net torque: mean (standard error).

		Moment of inertia condition						
Variable	NBP	1	2	3	4	5	6	7
T_{Net} (N·m)	583.56* (9.15)	$840.32^{E,F}$ (11.55)	803.95^{F} (9.87)	875.89^{E} (13.23)	1165.62^{D} (22.85)	1314.51^{C} (17.14)	1530.73^{B} (21.52)	1730.64^{A} (24.40)

* Significant differences found between loaded and unloaded walking conditions.
Within MOI manipulation, conditions with same letter are not significantly different.

In the present study, continuous and discrete relative phase are measures of the amount of counter-rotation between the pelvis and thorax during walking. Significantly smaller values for both continuous and discrete relative phase (Table 7) were found in the BP condition compared to the NBP condition. A significant main effect of MOI was found for ϕ_{cont}, but not for ϕ_{disc}. Increasing MOI resulted in larger values for ϕ_{cont}, indicating a more out-of-phase relationship between pelvic and thoracic rotation.

Table 7. Relative phase: mean (standard error).

		Moment of inertia condition						
Variable	NBP	1	2	3	4	5	6	7
ϕ_{cont} (Deg)	115.76* (0.96)	68.40^{E} (1.54)	69.39^{E} (1.77)	71.51^{E} (1.73)	79.61^{D} (1.73)	86.18^{C} (1.79)	89.71^{B} (1.72)	97.13^{A} (1.62)
ϕ_{disc} (Deg)	123.20* (1.62)	61.13 (2.13)	65.89 (2.35)	63.62 (2.18)	67.92 (2.18)	66.88 (2.23)	67.81 (2.16)	72.61 (2.20)

* Significant differences found between loaded and unloaded walking conditions.
Within MOI manipulation, conditions with same letter are not significantly different.

Carrying a BP resulted in a significant increase in oxygen consumption, oxygen consumption relative to subject's body mass; and oxygen consumption per total mass carried (Table 8) compared to unloaded walking. No significant main effect of MOI was found on oxygen consumption (VO_2, VO_2/BM, VO_2/TM).

Table 8. Oxygen consumption: mean (standard error).

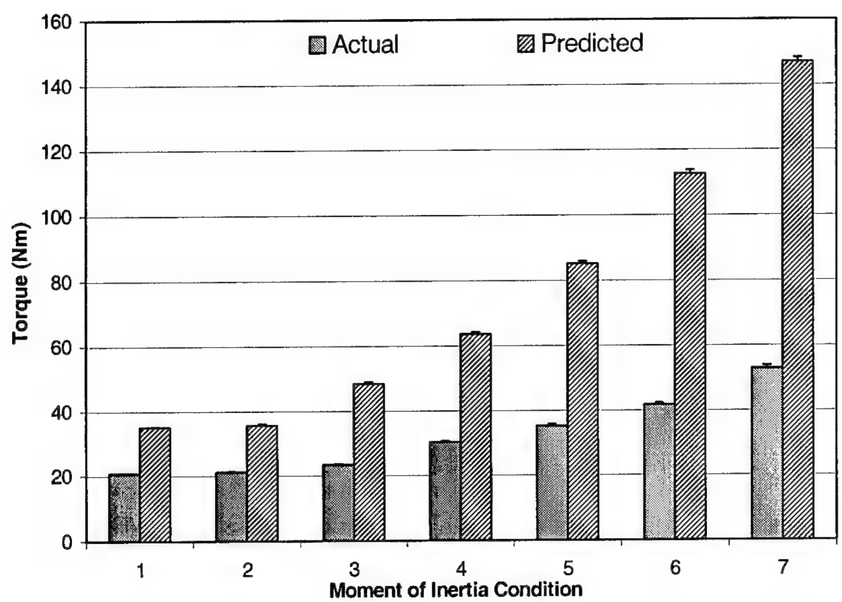
		Moment of inertia condition						
Variable	NBP	1	2	3	4	5	6	7
VO_2 (mL/min)	903.30* (16.76)	1443.50 (25.88)	1452.58 (24.91)	1442.52 (28.57)	1472.65 (29.57)	1475.17 (25.28)	1515.28 (26.63)	1498.69 (25.96)
VO_2/BM (mL·kg·min)	13.50* (0.25)	21.45 (0.29)	21.60 (0.29)	21.41 (0.32)	21.82 (0.31)	21.91 (0.27)	22.54 (0.32)	22.27 (0.29)
VO_2/TM (mL·kg·min)	13.50* (0.25)	14.90 (0.20)	15.00 (0.20)	14.87 (0.23)	15.15 (0.22)	15.21 (0.19)	15.66 (0.22)	15.46 (0.20)

*Significant differences found between loaded and unloaded walking conditions.

DISCUSSION

The results of this study support the hypothesis that the transverse plane upper body torque increases as a result of systematic increases in the BP MOI during walking at a constant walking speed (1.3 ms^{-1}). However, this increase in upper body torque was less than predicted solely from the increase in MOI as a result of changes in load distribution in a BP (Figure 3). As expected, the lower than predicted upper body torque was associated with systematic decreases in the acceleration of the upper body as MOI increased. These results are consistent with the previous study (10) in which the speed of loaded walking was manipulated. The findings add further support to the notion that the potential effects of increasing MOI on upper body torque are attenuated during load carriage. The purpose of this study was to further investigate the adaptations in gait that attenuate upper body torque while carrying a load.

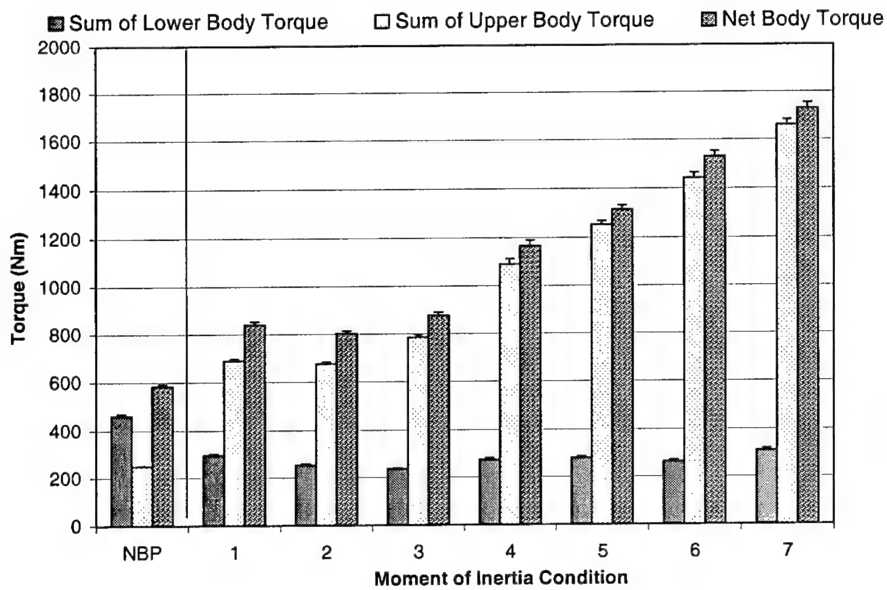
Figure 3. Actual and predicted peak upper body torque



On the basis of the findings of the study by LaFiandra et al. (10), it was hypothesized that if a goal of load carriage is to prevent the generation of upper body torque, the lower body torque would decrease with increasing upper body MOI so that the potential transmission of torque to the upper body would be diminished. However, although the lower body torque was reduced compared to the NBP condition, and was relatively small (about 15% of the upper body torque at the highest MOI condition), it remained constant across all MOI conditions. The constant lower body torque across conditions also failed to support the alternative hypothesis that the lower body torque would increase as the need to counteract the increasing upper body torque. Additionally, the upper body torque accounted for close to 100% of the net body torque (Figure 4), indicating that the lower body torque had little effect on net body torque. Our

interpretation of this result is that the large upper body MOI presented the system with a mass that was reluctant to changes in rotation. Thus, a further reduction in the lower body torque as the upper body MOI increased was unnecessary to prevent torque transmission to the upper body.

Figure 4. Comparison of net body torque, sum of upper body torque and sum of lower body torque



It was additionally hypothesized that as upper body MOI increased, a further decrease in counter-rotation (out-of-phase relation) would be observed to minimize the transmission of torque to the upper body. The results did not support this hypothesis. A small but significant increase in the phase relation was observed across increasing MOI conditions (from 68° to 97°). This finding would at first seem to support the alternative hypothesis that the counter-rotation would increase in order to counterbalance the effects of increasing MOI. However, since the magnitude of upper body torque is far greater than that of lower body torque, the lower body torque has negligible influence on the net body torque. Therefore, we would argue that the increased counter-rotation is negligible in light of the greater resistance to changes in rotation due to the increased MOI.

Overall, the idea that there are adaptations in gait that occur to either prevent or counterbalance large increases in upper body torque explain only to a limited extent the findings of the present study. The reluctance of the upper body changes in rotation in the large MOI conditions suggests a relaxation of the adaptations required to prevent or counteract the potentially large upper body torque. The increased inertial constraint of the upper body in the high MOI conditions may actually release movement degrees of freedom. For example, the observed increase in out-of-phase relation between upper and lower body at higher MOI conditions is closer to that observed in unloaded walking at the same speed. Similarly, further decreases in lower body torque are unnecessary, potentially allowing for the trend to greater transverse plane pelvic rotation at the

highest MOI condition and, thus, longer stride lengths (7). In future studies, we plan to test this idea by performing a similar experiment in which the walking task requires higher speeds.

Results of the analysis of metabolic cost add further support to the proposal that the other segments may not need to play an active role in controlling upper body torque as MOI increases. While there were significant increases in oxygen consumption between NBP and loaded BP, there were no further increases as a function of increases in MOI. The increase in upper body torque in the high BP MOI conditions apparently elicits minimal increases in muscular force to control the load.

One variable that was not considered in this experiment was the potential influence of intersegmental stiffness on the transmission of forces between the upper and lower body. Increased stiffness between segments would facilitate torque transmission, while a decrease in stiffness would reduce transfers. We have previously argued the dynamic relationship between the upper and lower body during gait may be modeled as two segments connected by a torsional spring (10), described by Equation 4:

$$I\ddot{\theta} = k\theta \qquad (4)$$

$I\ddot{\theta}$ represents the torque stored in the spring, k, the stiffness of the torsional spring, and θ, the amplitude of the angle between the two segments. In physiological systems, stiffness is mediated by co-contraction of antagonistic muscle pairs; in this case, the internal and external abdominal obliques. The more out-of-phase pattern of pelvic and thoracic rotation in the large BP MOI conditions suggests an increase in angular displacement, θ, and assuming constant torsional stiffness, an increase in the amount of torque transmitted between the upper and lower body. However, the torsional spring model illustrates that decreasing torsional stiffness, k, is an alternative strategy for decreasing the amount of torque transmitted between the segments and, in turn, decreasing lower body torque. Research is currently underway to explore the role of torsional stiffness on torque transmission during load carriage.

In summary, the idea of a relaxation of the adaptations in gait that minimize the potential effects of a large MOI is consistent with the views of Nicolai Bernstein (1) who stated "The secret of coordination lies not in wasting superfluous force in extinguishing reactive phenomena but, on the contrary, in employing the latter in such a way as to employ active muscle forces only in the capacity of complementary forces. In this case the same movement (in the final analysis) demands less expenditure of active force."

CONCLUSIONS

Systematic increases in the MOI of the BP during walking resulted in an increase in upper body torque, but the increase was less than predicted solely from the increase in MOI as a result of changes in load distribution in a BP. The lower than predicted upper body torque was associated with systematic decreases in acceleration of the upper body as MOI increased.

While lower body torque was reduced and remained relatively small compared to the NBP condition, lower body torque values remained constant across all MOI conditions. In addition, with the upper body torque accounting for close to 100% of the net body torque, it was concluded that the lower body had little effect on net body torque.

The large upper body MOI presented a system that was reluctant to changes in rotation, and a further reduction in the lower body torque as the upper body MOI increased was unnecessary to prevent torque transmission to the upper body.

Increasing upper body MOI did not result in a more in-phase pattern of pelvic and thoracic rotation, but a significant increase in the phase relation with increasing MOI conditions was found. Considering the lower body torque has negligible influence on net body torque, the increased counter-rotation is negligible in light of the greater resistance to changes in rotation due to the increased MOI.

The reluctance of the upper body changes in rotation in the large MOI conditions suggests a relaxation of the adaptations required to prevent or counteract the potentially larger upper body torque. The increased inertial constraint of the upper body in the high MOI conditions may actually release movement degrees of freedom. Further decreases in lower body torque are unnecessary, potentially allowing for the trend to greater transverse plan pelvic rotation at the highest MOI condition and, thus, create longer stride lengths.

With no significant changes in metabolic cost with increasing upper body MOI, it was concluded that the increase in upper body torque in the high BP MOI conditions apparently elicits minimal increases in muscular force to control the load.

REFERENCES

1. Bernstein, N. *The Coordination and Regulation of Movement.* London: Pergamon Press, 1967.
2. Borhese, N. A., L. Bianchi, and F. Lacghaniti. Kinematic determinants of human locomotion. *J. Physiol.* 494: 863-879, 1996.
3. Dempster, W. T. *Space Requirements of the Seated Operator.* Wright Paterson Air Force Base. Technical Report WADC-TR-55-159, 1955.
4. Harman, E. A., K. H. Han, P. Frykman, and C. Pandorf. *The Effects of Backpack Weight on the Biomechanics of Load Carriage.* Natick, MA: USARIEM. Technical Report No. T00/17, 2000.
5. Inman, V.T., H.J. Ralston, and F. Todd. *Human Walking.* Baltimore, MD: Williams and Wilkins, 1981.
6. Kinoshita, H. Effects of different load and carrying systems on selected biomechanical parameters describing walking gait. *Ergonomics* 28: 1347-1362, 1985.
7. LaFiandra, M.E., K. G. Holt, R. C. Wagenaar, and J. Obusek. Transverse plane kinetics during treadmill walking with and without a load. *J. Clin. Biomech.* 17(2): 116-122, 2002.
8. LaFiandra, M. E., R. C. Wagenaar, K. G. Holt, and J. Obusek. Mechanisms of changes in kinematic patterns during load carriage in walking. *J. of Biomech,* (In Press).
9. Obusek, J. P., E. A. Harman, P. Frykman, C. J. Palmer, and R. K. Bills. The relationship of backpack center of mass location to the metabolic cost of load carriage. *Official Journal of the American College of Sports Medicine* 29: S205, 1997.
10. Patton, J. F., J. Kaszuba, R. P. Mello, and K. Reynolds. Physiological responses to prolonged treadmill walking with external loads. *Eur. J. Appl. Physiol.* 63: 89-93, 1991.
11. Pierrynowski, M. R., R. W. Norman, and D. A. Winter. Mechanical energy analyses of the human during load carriage on a treadmill. *Ergonomics* 24: 1-14, 1981.
12. Portney, L. G., and M. P. Watkins. *Foundations of Clinical Research Applications to Practice, 1st edition.* Norwalk, CT: Appleton and Lange, 1993.
13. Queseda, P. M., L. J. Mengelkoch, R. C. Hale, and S. R. Simon. Biomechanical and metabolic effects of varying backpack loads on simulated marching. *Ergonomics* 43: 293-309, 2000.

14. Soule, R. G., K. B. Pandolf, and R. F. Goldman. Energy expenditure of heavy load carriage. *Ergonomics* 21: 373-381, 1978.
15. Tilbury-Davis, D. C., and R. H. Hooper. The kinetic and kinematic effects of increased load carriage upon the lower limb. *Hum. Mov. Sci.* 18: 693-700, 1999.
16. Townsend, M. A. Dynamics and coordination of torso motions in human locomotion. *J. Biomech.* 14: 727-738, 1981.
17. van Emmerik, R. E. A., and R. C. Wagenaar. Effects of walking velocity on relative phase dynamics in the trunk in human walking. *J Biomech.* 29: 1175-1184, 1996.
18. Wagenaar, R. C., and W. J. Beek. Hemiplegic gait: A kinematic analysis using walking speed as a basis. *J. Biomech.* 25: 1007-1015, 1992.
19. Wagenaar, R.C., and R.E.A. van Emmerik. Resonant frequencies of arms and legs identify different walking patters. *J. Biomech.* 33: 835-861, 2000.